AN APPROACH TO MODEL THE HEAD – BICYCLE HELMET DYNAMIC BEHAVIOUR THROUGH TRANSIENT FINITE ELEMENT ANALYSIS

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Abstract:

This paper focuses on the mechanical characterisation of the helmets by using the finite element method as a first step in the helmet optimisation and improvement process. Considering the higher risk for head injury in traffic accidents, it is of crucial importance to be able to provide relevant and adequate head protection. The relation between different parameters (materials properties, vents number and localization, etc.) and the risk of head injury is evaluated. The finite element software that is used is MSC Marc Mentat. Starting from STL (standard triangulation language) files different finite elements models were built to compare the mechanical behaviour of the different types of cyclist helmets during the impact. The linear acceleration and the energy absorption were evaluated for a better understanding of the mechanism through which the helmet protects the head during an impact. All materials studied here protect the head by reducing the linear acceleration by more than 80% and stress values into the head by more than 65%. The anisotropic foam studied in this paper seems to offer the best protection against linear acceleration, stress and deformation. The study shows promising results for the use of anisotropic foams in the process of improving the head protection offered by a helmet.

Introduction

Cycling represents one of the most popular recreational sports, but also one of the main transportation means all over the world. In 2000, road traffic injuries caused the death of an estimated 1,200,000 humans, representing the ninth highest cause of death worldwide (United Nations, 2003). Many previous studies have outlined the effectiveness of using bicycle helmets (Thompson, 1989; Thomas, 1994) and also the possibility of the improvement of currently used helmets design (McIntosh, 1995; Ching, 1997). From a survey in 1999, in Belgium, 9.6% of the study subjects took a ride on their bikes. From this percentage 7.3% used the bike as their main method of transportation to work and 19.2% to school (NIS, 2000). In 2000, a total of 6655 injuries and 134 fatalities were attributed to cycling accidents in Belgium on public roads. 9.8% of all road accidents and 9.1% of road traffic deaths were represented by pedal cyclists (NIS, 2000). However, many of these accidents would have been prevented.

The head represents the most vulnerable part of the human body and the most important to protect. Between 21 to 61% of the victims of bicycle accidents seeking medical care has a head injury (Collins, 1993; Eilert-Petersson, 1997; Fife, 1983; Wood, 1988). The medical care requested by this increased number of victims implies high costs all over the world. Bicycling also tragically leads to increased numbers of fatalities due to head injuries. Head injuries represent the cause of death in 69–93% of fatal bicycle accidents (Fife, 1983; Guichon, 1975; Öström, 1993; Wood, 1988).

The bicycle helmet protects the heads against an impact by reducing the impact energy that is transmitted to the head through energy absorption and dissipation by elastic and plastic deformation of its components (the foam liner and the outer shell) and by increasing the area over which the impact is distributed. Currently, the protective quality of the helmet is most often evaluated through the Head Injury Criterion (HIC). HIC (Mellander, 1986) summarises the relationship between linear acceleration, impact duration and the onset of skull fractures (Delye, 2007). However, in reality the patterns of impact, stress and tissue deformation encompass a much higher degree of complexity than this relatively simple criterion would indicate. The variant nature of all the implied factors and the complex relationship between them, complicate the process of defining one single mechanism for brain injury as a result of head impact, or to associate a given degree of injury with a given type of impact force. Moreover, the tolerance criterion, which relates the occurrence of all brain injuries exclusively to translational head motions, has been criticized for a long time (Hirsch, 1970). Therefore, precise lesion-specific tolerance criteria for head injury are needed. By combining these characteristics of specificity with the relative frequencies of different traumatic brain lesions, a systematic approach for the improvement of helmet protectiveness can be described.

The rationale of this study is that in order to improve the helmet, an initial better understanding of the helmet's mechanical role and biomechanical behaviour of the head during an impact is essential. Therefore, the mechanisms, according to which the current helmets transmit the impact energy to the head, are investigated.

The use of new designs, new materials for the foam liner with more adequate mechanical properties may improve the degree of protection offered by the helmet against normal and tangential forces. Previous research work in our group indicates the use of anisotropic foams as appropriate for increasing the head protection offered by a helmet during a bicycle accident (Depreitere, 2004; Depreitere, 2007; Van Lierde, 2005). The results obtained from previous studies underline the quality of this foam in reducing the rotational acceleration when subjecting to an angular impact, but an analysis of linear acceleration reduction is also necessary.

This paper evaluates the influence of using different material properties for the foam liner and different geometrical parameters on the helmet's protective capability in terms of energy absorption and protection against linear acceleration under different impact conditions.

Materials and methods

Finite element models design

The foam model was designed starting from an STL file of an adult bicycle helmet with air vents, using Mimics® and Magics® (Materialise, Haasrode, Belgium). The used helmet design, known as "child type", assures a better coverage of the head temporal area (Depreitere, 2004). To create another solid helmet version, the vents were covered using various Boolean operations via Magics® software. The helmet model is composed of an internal foam liner of ~20 mm, an external shell of 0.5 mm and internal soft comforts pads.

In order to obtain a realistic position of the head into the helmet, the helmet STL and the head STL were imported into Magics®. By translational and rotational movements, the sagittal plane of the head was aligned to the sagittal plane of the helmet.

By combining the two foam helmet models with the head form, two FE models were built:

– model 1 – helmet with vents (figure 1 upper row);
– model 2 – helmet without vents (figure 1 lower row).

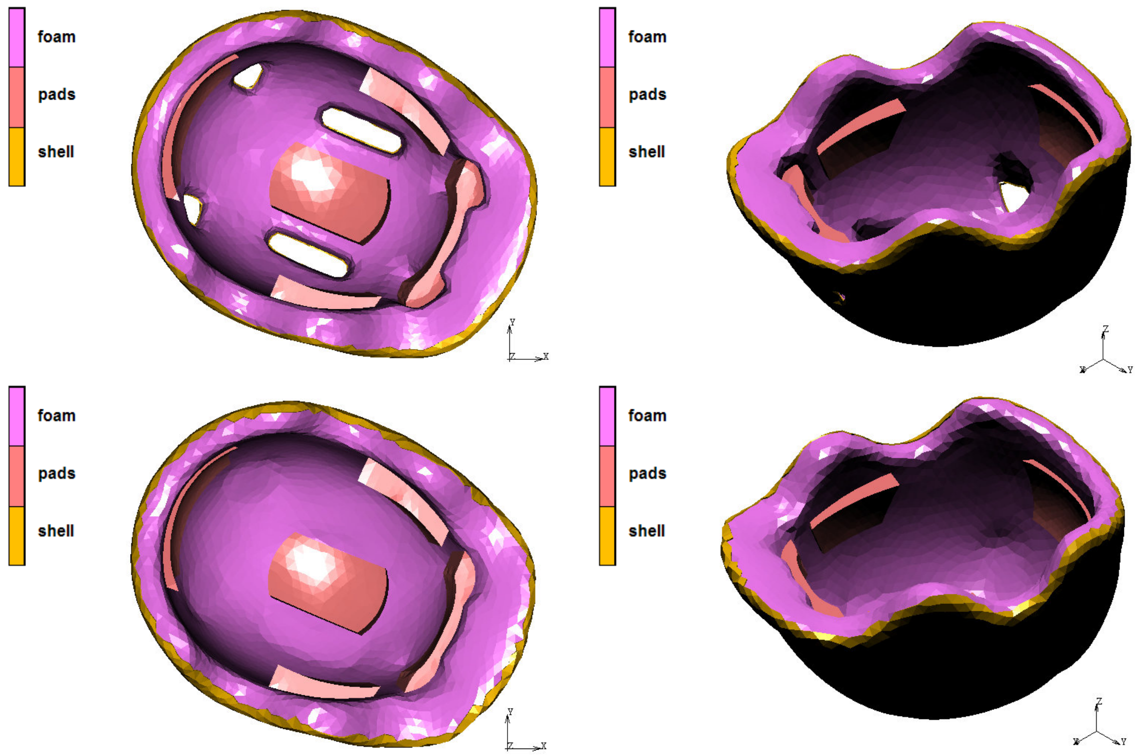

Fig. 1: ***The child type helmet models*** *– with (upper row) and without vents (lower row)*

The head model was created using the same software, starting from the STL file of an adult dummy head. The head form was modelled as a single isotropic purely elastic body (figure 2).

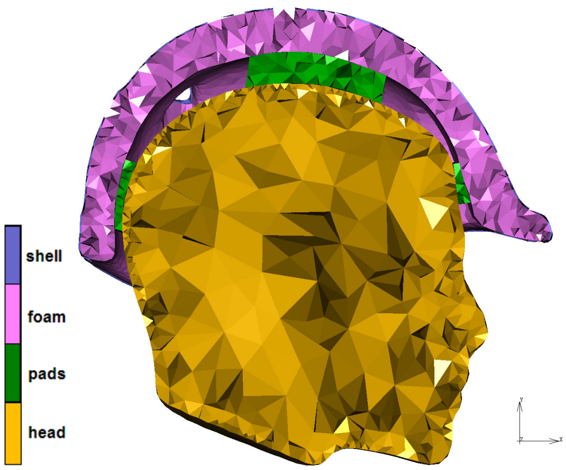

Fig. 2: ***Final model (outer shell, foam, comfort pads, head form)***

All of the models' components were then meshed in MSC.Patran® using 4-node linear isoparametric three-dimensional tetrahedron elements with ~6 mm edge size. After obtaining the solid mesh, the models were imported in MSC Marc Mentat for the next two phases: the processing and postprocessing phases. The external shells for both of the helmets were created in MSC Marc by duplicating and by applying a uniform offset of 0.25mm to the external surface of the foam liner. By copying the same geometry it was possible to achieve an intimate contact between the shell and the foam liner. A uniform shell thickness of 0.5mm was obtained.

The analyses were run using MSC.MARC/Mentat® software. No friction between the existing interfaces was considered in this study, so an unrestrained sliding was allowed.

Impact simulations – the influence of material properties

Linear elastic and isotropic material properties were assigned for the shell (polycarbonate PC) and head model components (standard crash dummy head - Duraluminium material properties) (table 1). A real mass of 4.5 kg for the adult head was obtained.

Previous data obtained after impact tests in non-perpendicular angles in this multidisciplinary bicycle helmet research program have shown promising results for using anisotropic foams for the purpose of reducing rotational acceleration (Depreitere, 2004; Van Lierde, 2005). For evaluating the influence of material properties on the linear energy absorption ability and for characterizing the mechanical behaviour of the anisotropic polyether sulphone foam (PES) during an impact, a comparison between PES and three different types of expanded polystyrene was made. The anisotropic foam material properties were obtained by experimental testing (see table 1). The expanded polystyrene (EPS) foams were considered to have isotropic pure elastic material behaviour and the materials properties were taken from literature (Zhang, 2004).

Table 1: ***Material properties used in the finite element models***

Material properties / **Type of material**	**Young Modulus (E) (MPa)**		**Shear Modulus (G) (MPa)**		**Poisson ratio (υ)**		**Density (ρ) (g/mm^3)**
Polycarbonate	2000		-		0.37		$1.12 \cdot 10^{-3}$
Head	2500		-		0.3		$1.247 \cdot 10^{-3}$
EPS high density	19.88		-		0.17		$58.15 \cdot 10^{-6}$
EPS medium high density	15.75		-		0.14		$50.98 \cdot 10^{-6}$
EPS medium density	10.05		-		0.11		$39.85 \cdot 10^{-6}$
EPS low density	2.5		-		0.08		$24.3 \cdot 10^{-6}$
PES	$\mathbf{E_1}$	**0.55**	$\mathbf{G_1}$	**0.5**	υ_1	**0.1**	$\mathbf{57 \cdot 10^{-6}}$
	$\mathbf{E_2}$	**0.55**	$\mathbf{G_2}$	**3**	υ_2	**0.1**	
	$\mathbf{E_3}$	**23**	$\mathbf{G_3}$	**3**	υ_3	**0.1**	

The anisotropic PES was assumed to have an elastic-plastic transversally isotropic material behaviour. The yielding of the foam was assumed to occur above a stress value of 0.56 MPa. Due to its macroscopic cellular structure we assume that the cells are always perpendicular to the head surface. This assumption resulted in two principal material orientations: one along the longitudinal axis of the cells and the other in a perpendicular plan (figure 3).

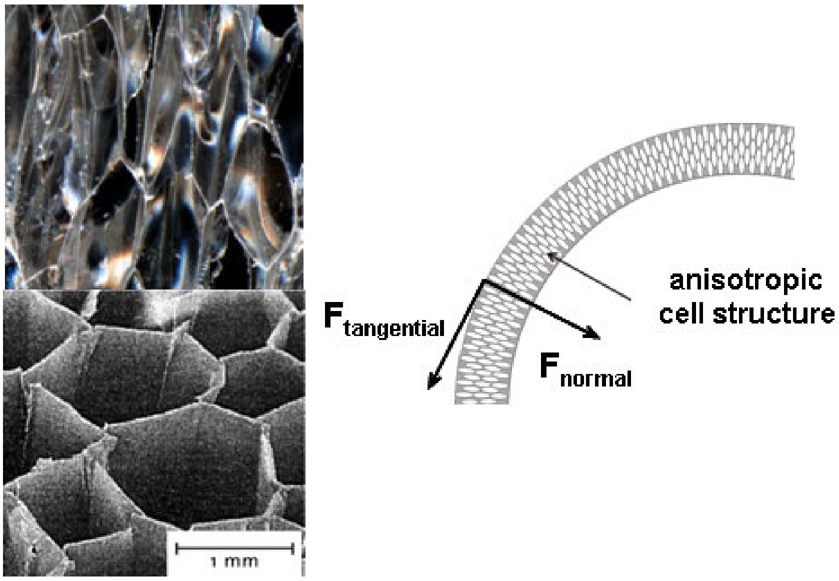

Fig. 3: ***The PES anisotropic foam cellular structure***
- Left pictures – microscopic view of longitudinal section (upper picture) and transversal section (lower picture)
- Right figure – schematic representation of the cellular structure

For achieving the permanent connection between the head and the helmet, two pairs of elastic springs with a damping coefficient of 281 and a stiffness coefficient of 10 were added to the models.

Fig. 4: ***Impact on the vertex***

The finite element simulation analyzed a vertical drop of the assembly on a rigid surface from a distance of ~8 mm (figure 4). An initial velocity of 5.23 mm/s was applied to all the bodies so that an impact velocity of ~20 km/h was obtained. This value corresponds to a value for the impact energy of ~75J. A total time of 15ms was considered for the simulation so that only short impact durations were allowed. The impact was localized, in this case, at the vertex.

Results

When the helmet impacts the rigid surface, only few nodes of the elastic shell initially come into contact. Soon these nodes gain a velocity in the opposite direction while still maintaining the contact with the rigid surface. This may cause an initial oscillation of the contact force, immediately after the impact occurs. The head impact takes place after ~0.08s from the beginning of the simulation. After the head touches the internal surface of the foam layer, the nodes of the shell, which are in contact with the rigid surface, again acquire velocity in the direction of the impact. As a result, the contact force between the shell and rigid surface increases. The contact force increases and reaches a maximum. After that, the helmet starts rebounding. The head and helmet separate again and the contact force at this interface drops. The displacement of head and helmet during the impact is plotted in figure 5.

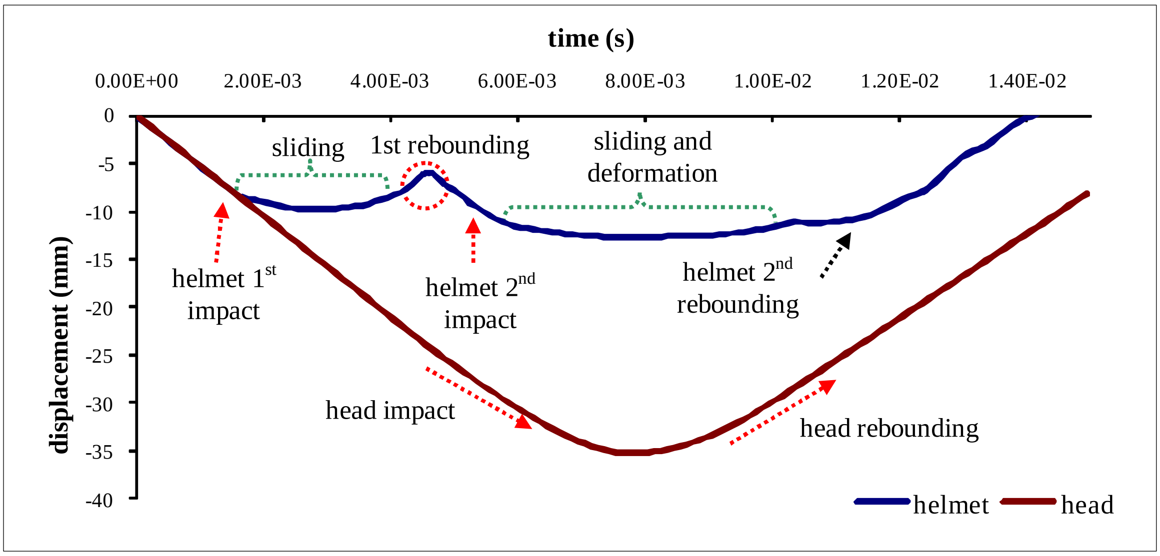

Fig. 5: ***The displacement of the PES helmet and head along the impact direction***

During the impact, the helmet develops a visible vibrational behavior. Moreover, the main plastic deformation of the foam occurs during the final sliding phase. The foam thickness reduction due to the deformation is ~80% in the impact zone (figure 6).

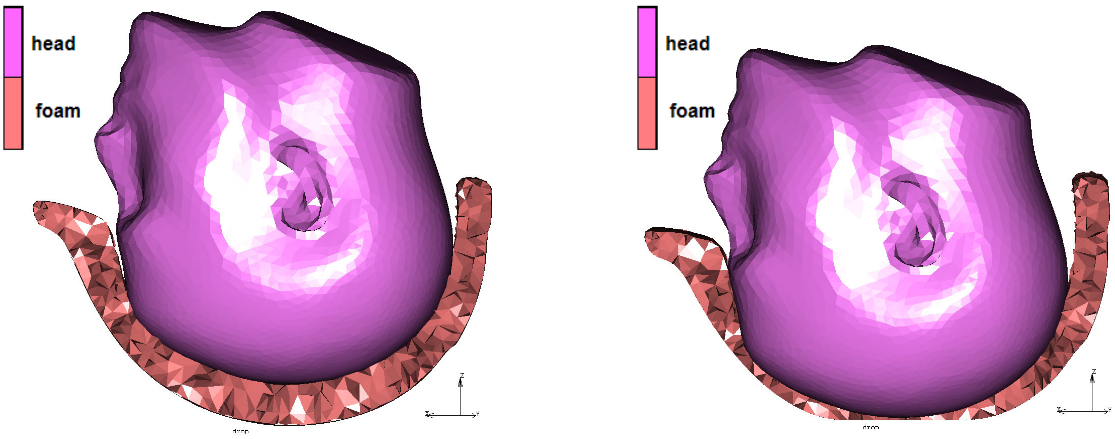

Fig. 6: ***Solid views showing the foam layer deformation caused by the impact (right) comparing with the undeformed situation, before the impact occur (left)***

By this plastic deformation mechanism, all foam materials studied here reduce the head stress by more than 65 % (figure 7). The PES foam shows a better behavior when considering the head protection by reducing the head stress by ~70%.

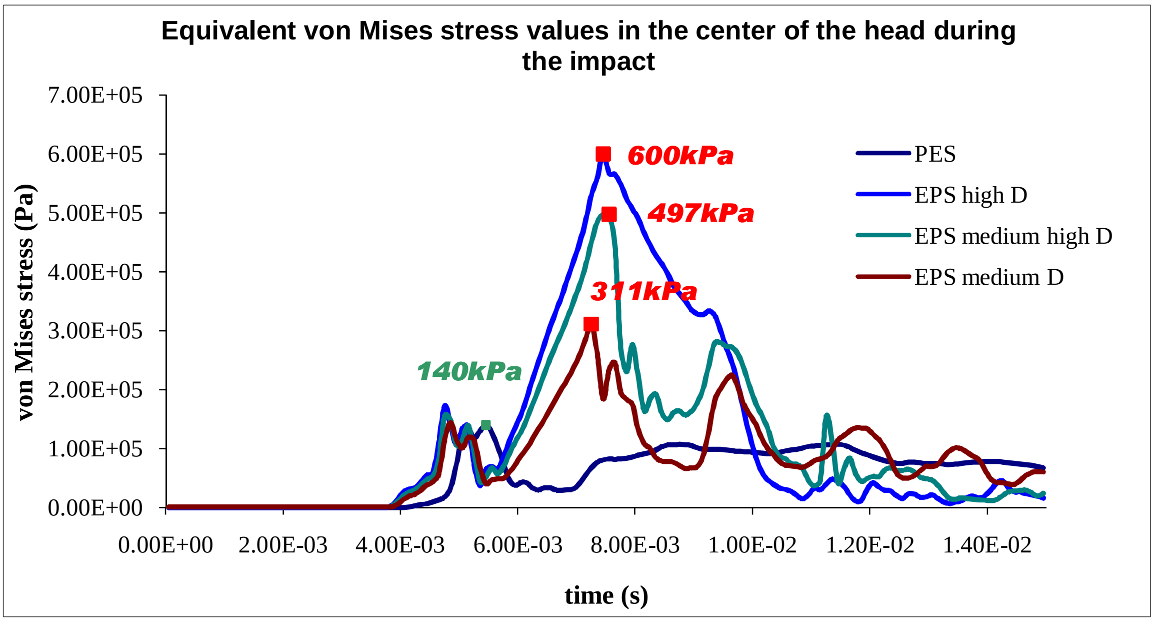

Fig. 7: ***The Von Mises stress evolution in the head center during the impact***

All helmets reduce the peak linear acceleration of the head during an impact by one order of magnitude. During the impact, there is more than 85% reduction in the maximum linear acceleration of the head when comparing all types of EPS studied here (figure 8).

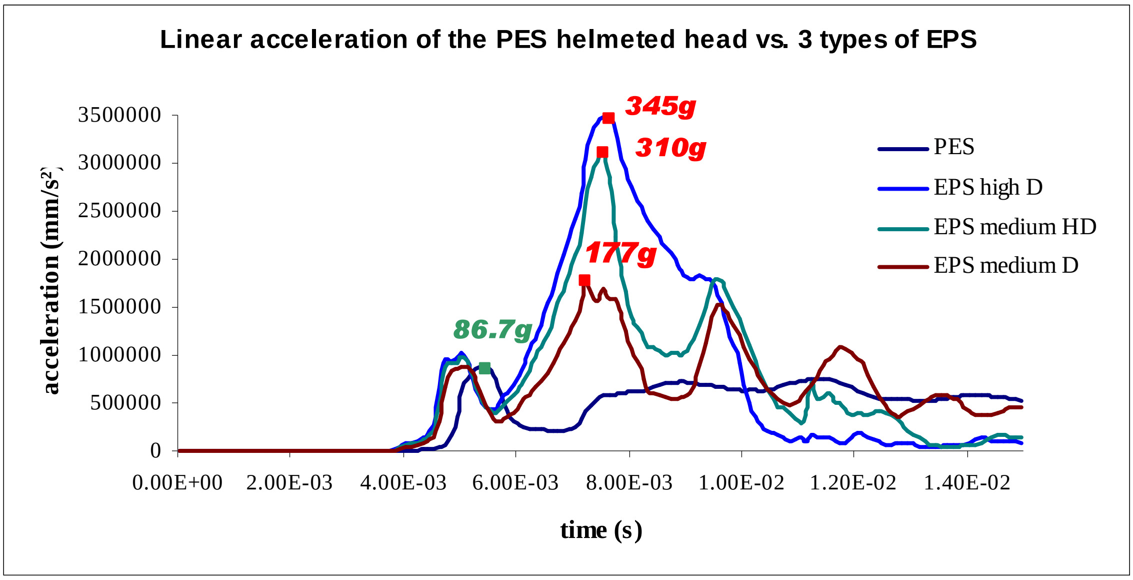

Fig. 8: ***The linear acceleration of the head in the case of PES helmet use in comparison with the linear acceleration of three types of EPS helmets during an impact***

In the case of high density and medium-high density EPS, the resulting linear acceleration still remains above the peak head linear acceleration tolerance limit considered in the several governmental helmet standards (figure 8) (300 g for CPSC and Snell B95). This indicates that the risk for head injury is still significant using these materials. From all types of EPS studied the most important reduction in linear acceleration is achieved when EPS medium density foam is used.

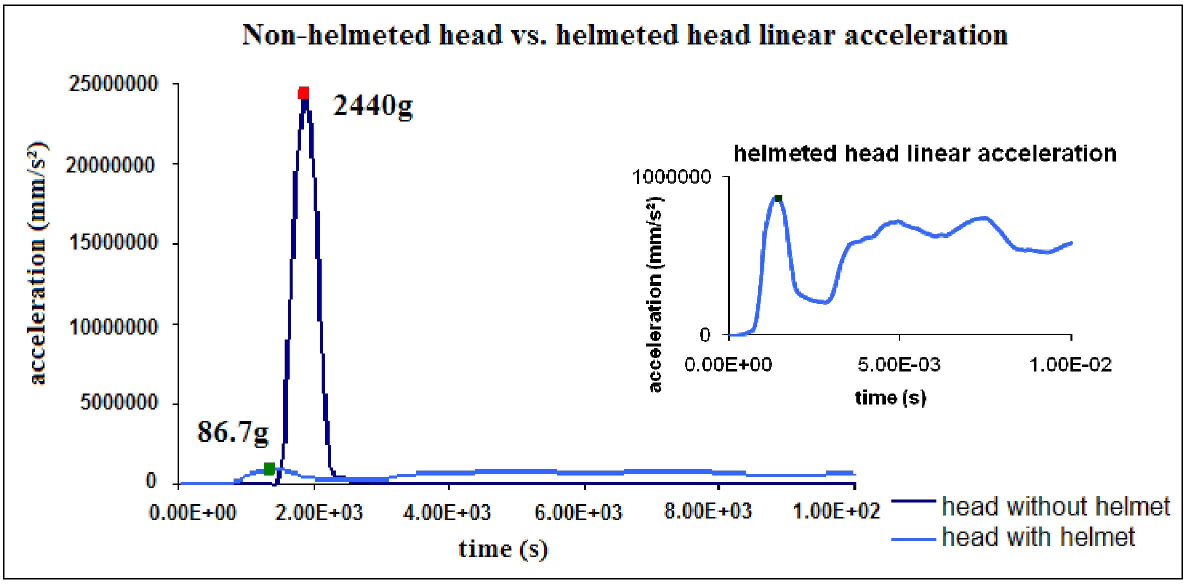

Fig. 9: ***The linear acceleration of a non helmeted head in comparison with a PES helmeted head during an impact (on the impact direction)***

The reduction is more significant for the case of PES when comparing with all cases of expanded polystyrene studied here (figure 9). The use of PES helmet determine a reduction of head linear acceleration by ~95 % when comparing the peak value obtained during the impact with the peak value obtained in the case of a non-helmeted head impact (figure9).

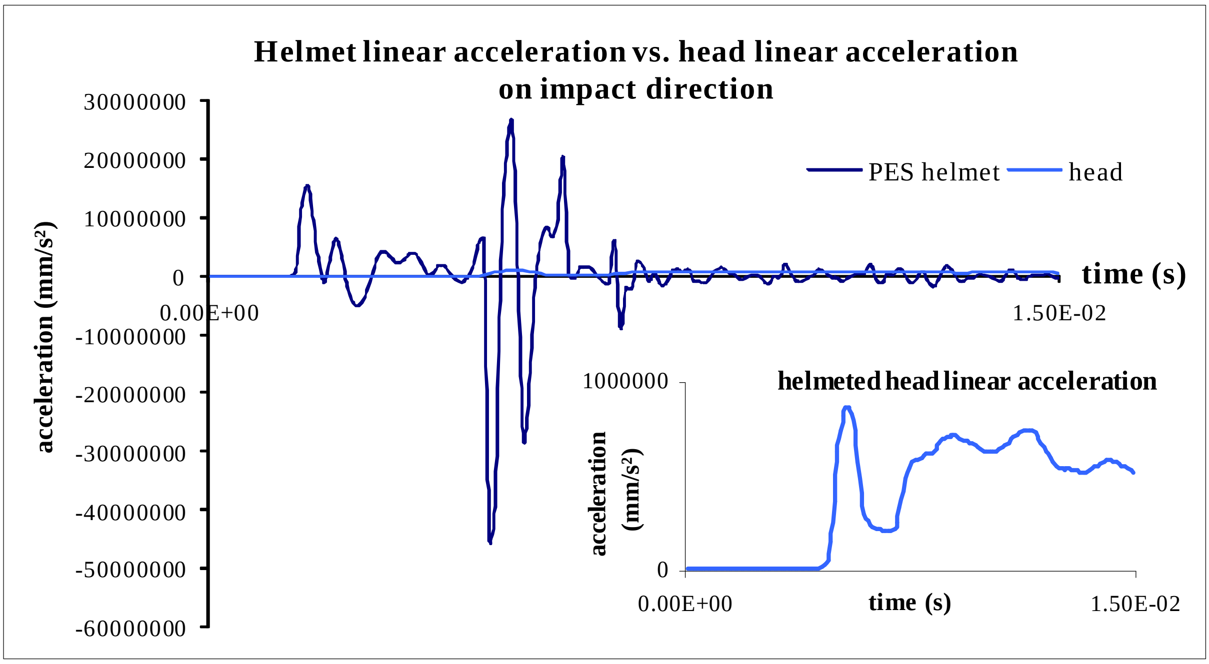

Fig. 10: ***The linear acceleration of the head in comparison with the linear acceleration of the PES helmet during an impact (on the impact direction)***

The influence of wearing the helmet on the head linear acceleration is described also in figure 10 by plotting the linear acceleration of the helmet in comparison with the head linear acceleration during the impact (figure 10).

For the case of low density EPS, the simulation became instable before reaching the peak impact force. For this reason, the results are not included in the comparisons presented here. This instability may be caused by an extreme deformation in compression for the low density EPS during the impact due to the very low Young modulus assigned to the foam material.

Discussion and conclusions

This study describes the protective mechanisms of different types of helmets in vertical impact conditions. Also, the influence of the foam material properties on the process of impact energy transmission to the head is observed. In this study, the head is considered as being a uniformly deformable body so that an external map of the stress distribution and evolution during an impact can be obtained. The contact forces and stresses following an impact for the helmeted head are compared with those of a non-helmeted head.

The study shows a superior material behavior for the PES foam (polyether sulfone foam) when comparing with different expanded polystyrene foam densities for both of the protective aspects studied here (energy absorption and linear acceleration). The results presented in this paper are comparable with values from literature (Deck, 2003).
Bicycle helmets are primarily designed to reduce the effect of linear forces by providing a soft crushing layer, which reduces the peak linear acceleration of the head and brain during impact. The current standards specify that the peak forces of acceleration shall not exceed 300g (300 g for Consumer Product Safety Commission – CPSC Standard and Snell B95 standard) (Bicycle Helmet Safety Institute, 2008). The results of the study of McIntosh (McIntosh et al. 1995) showed that even a peak linear acceleration below 300g can correspond to higher HIC-values. Based on that study, the peak linear acceleration threshold in the Australian Standard was lowered from 300 g to 250 g (250 g for EN 1078 and EN 1080) (UK Department for Transport, 2008), although the author had proposed a threshold of 200 g. Our study shows that for two of the tested EPS foams the linear acceleration values are above the limit of 300g established in the McIntosh study. For the PES anisotropic foam the observed linear acceleration is significantly lower compared with the same standards, indicating an increased head protection.

We have to consider that no helmet can protect against all possible impacts, and that death or serious injury could happen in extreme circumstances. For providing a maximum protection, the helmet must fit the rider's head properly, and the rider must properly put the helmet on and hook the straps together, following the proper instructions.

Moreover, head impacts from bicycle crashes do not generally involve a direct perpendicular impact with its vector intersecting the center of rotation of the head. Most commonly, there is an angled impact as the head hits the ground with forward momentum. Such an impact is likely to impose some degree of rotational force on the head and brain. Contact forces, accelerations and locally developed shear strains can cause the principal neuropathological features of head injury. Contact forces resulting from an object striking the head produce local effects (scalp laceration, skull fracture, extradural haemorrhage) and also, acceleration induces relative movement of the brain in the skull, which leads to intracranial and intracerebral pressure gradients. Acceleration, both linear and rotational, can result in functional injury of the brain even in the absence of a skull fracture. The shear strains can cause contusion of the brain and tearing of blood vessels. The extent to which the brain tissue is internally deformed depends on the location of the impact point, the distribution of the forces and the resulting motion of the head. But the mechanisms causing head injuries are still not fully understood. Usually, individual mechanisms of trauma produce very specific types of head injuries (Youmans, 1996). The injury pattern depends not only on the mechanical aspects but also on the complex interaction of the subsequent pathophysiological events.

Our future research will focus on the PES protective effect against rotational acceleration under angular impacts: indeed previous studies have shown an improved biomechanical behavior for this type of anisotropic foam (Depreitere, 2004). It is noteworthy to mention that the rotational forces may be reduced by virtue of the same crushing effect of the helmet, which is proven to reduce linear forces. On the other hand, the helmet presence may increase the rotational forces by increasing the size and mass of the head. The effect of helmets on rotational forces affecting the brain is not entirely clear and it is not the main subject of this paper but represent thus an essential direction for further continuation of the research. It should be noted that there are presently no helmet safety standards that regulate the ability of a helmet in reducing the angular acceleration.

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